DEVELOPMENT OF THOR NT: ENHANCEMENT OF THOR ALPHA-THE NHTSA ADVANCED FRONTAL DUMMY

T. Shams, N. Rangarajan, J. McDonald, Y. Wang, G. Platten, C. Spade, P. Pope
GESAC, Inc.
United States
M. Haffner
HCS
United States
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ABSTRACT

The NHTSA Advanced Frontal Impact Dummy THOR has been enhanced to include improved anthropometry and biofidelity, durability, and ease of use. The previous version, known as THOR Alpha has been revised and is now called the THOR NT. The areas of improvement include improved anthropometry and biofidelity in the head with a single, integrated head skin which meets both isolated head drop and whole-body head impact requirements; more biofidelic dynamic response of the neck, including the introduction of an atlanto-occipital joint at the top of the neck with biofidelic range of motion; improved anthropometry at the shoulder and clavicles; improved biofidelity of the ribcage to the lower speed Kroell impact; improved anthropometry in the femur including a representation of the trochanter. The external forms for the pelvis and femur skin are now based on an undeformed shape to more correctly represent the interaction with a vehicle seat. More durable materials and improved production methods have been used, including injection molding the neck and the flexible joints in the spine. The pelvis and femur skins are made of PVC and the shoulders and front of the ribcage (bib) are made of more durable materials. The paper describes the enhancements in detail and presents the results from selected certification tests.

INTRODUCTION

Since the early 1980s, the National Highway Traffic Safety Administration (NHTSA) has supported the development of an advanced frontal crash test dummy with improved biofidelity under frontal impact conditions and with expanded injury assessment capabilities. This has involved extensive research in human anthropometry, biomechanics, and dummy development [Robbins, 1983; Schneider, 1985; Melvin, 1985; Schneider, 1992].

As part of the development effort, a prototype of a frontal crash test dummy, corresponding to a 50th percentile male, was completed in 1996. The principal features of the new crash dummy, known as THOR, have been described in White[1996] and Rangarajan [1998]. The dummy was tested at a number of different laboratories and the test results reported in a number of different proceedings [Ito, 1998; Hoofman, 1998; Petit, 1999; Shams, 1999]. During the initial testing process, a number of problems were identified. These were mainly in the area of poor durability of some of the components such as the neck and flexible joints in the spine, noise in some of the accelerometers, especially in the head and lumbar spine, and problems in handling and storage. Modifications were undertaken on the prototype THOR which resulted in the introduction of THOR Alpha. A description of the modifications is given in Haffner [2001]. The intent was that these modifications would address the principal issues that were raised by some of the laboratories. An extended series of tests were conducted by the Frontal Impact Dummy consortium in Europe, which included testing at TNO, INRETS, TRL, Bast, Insia, and the University of Heidelberg [ADRIA, 2000; Vezin,2002] . Additional testing was done by the NHTSA and GESAC at the University of Virginia and at the Applied Physics Laboratories. These tests indicated some of the important durability problems such as debonding of the neck, and the spine flex joints, tearing of the pelvis skin, shoulder pads, and bib were still occurring. In addition, the NHTSA asked GESAC to look at improving the anthropometry and biofidelity of the dummy where it was appropriate. The FID consortium provided the NHTSA with a detailed summary of their findings and along with comments from other labs and the improvements sought by the NHTSA, a list of potential modifications was arrived at.

THOR NT MODIFICATIONS

An extensive set of modifications were made to the THOR Alpha during the development of the THOR NT based on user comments and the need for improving the performance of the dummy. The

modifications could be roughly grouped according to the following:

- Anthropometry: Improvement in the external shape of selected segments, such as the head/neck interface, the shoulder, the pelvis, and the femur
- Durability: Improvement in the durability of many of the deformable parts of the dummy such as the neck, flex joints, and various stops, either through improved manufacturing process or redesign
- Usability: Improvement in the ease of use of the dummy, such as easier access to some instrumentation and providing a common electrical ground
- Biofidelity: Improvement in selected impact responses of the dummy, especially the thorax and head
- Fit and Finish: Improvement in the appearance of segments and the interfaces between segments

Head

The changes to the head included:

- Integrated head skin which covered both the skull and the face
- Improved chin anthropometry
- Zippered connection between head and neck skins
- Improved mounting of the nine-axis accelerometer package
- Modified mandible load plate
- Extended chin support
- Radiused edges on face plates

The most important change to the head was in going from separate head and face skins to a single, integrated head skin. It was also decided that the skin around the face should not possess any features, such as nose or lips, since a featureless face would improve repeatability during impacts.

The improved chin anthropometry was motivated by the need to improve the interaction with airbags if they happened to load the space between the chin and the neck. The bottom of the chin surface in the head skin was rounded to more closely model the 50th percentile male AATD chin. Figures 1 and 2 show the difference between the two head/face skin designs. The figures also show the improvement in the attachment of the neck skin to the head skin in the NT relative to the Alpha.
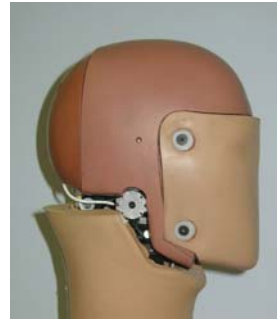

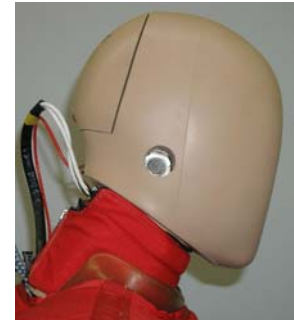Figure 1. THOR Alpha head/face.

Figure 2. THOR NT integrated head skin.

Also in the Alpha design, with a separable neck skin, it was sometimes possible for the airbag to intrude into the space between the neck skin and the head assembly, especially in out-of-position (OOP) situations. The zippered connection between the neck skin and the head skin was introduced to avoid that possibility. Also, the chin plate was extended downward to provide load support to the whole of the face, which was further helped by rounded Delrin extensions to the chin plate. The changes to the chin face plate also involved changing the geometry of the Confor foam used as the face response element in THOR. The edges of all the face plates were radiused to reduce the possibility of tearing into the head skin. Figures 3 and 4 show the differences between the two face plate designs.

Figure 3. THOR Alpha face plates.

Figure 4. THOR NT face plates.

The final modifications to the head involved improving the mounting of the nine-accelerometer package. In the Alpha, seven of the accelerometers were mounted on a special L-shaped block which mounted directly to the base of the head. The top two accelerometers were being mounted on a plug pressed into the top of the skull. In the NT, the mounting of

the top two accelerometers are now on a machined flat on the inside of the skull, which provides a more definite locating method for these two accelerometers relative to the C.G. accelerometers.

Neck

The changes in the neck included:

*   Using injection molding process to make the neck instead of bonding
*   Design of a new atlanto-occipital joint (A-O) which provides continuous resistance in flexion and extension
*   Improved neck spring load cells
*   Modification of the neck springs to include rubber inserts

The change to an injection molding process was motivated by the unreliability of the bonding method being used in the the THOR Alpha, where some necks failed early in the testing cycle. The new, molded necks are made of Butyl rubber instead of the earlier Neoprene, which provides some additional damping in the neck response. The injection molded neck is now extremely durable, repeatable, and reproducible. Up to 300 pendulum impacts were conducted on a single injection molded neck without failure. These tests were done with a pendulum pulse of 25G, but some overload tests up to 38G were also performed successfully.

The design of the A-O joint was changed to improve the response of the neck in the flexion/extension mode. The response at the A-O joint of THOR Alpha was controlled by two rubber stops which allowed for relatively free rotation of about 10° in flexion and 25° in extension. A drawing of the new cam and stop mechanism at the A-O is shown in Figure 5 and the response of the joint is compared with the original Alpha joint in Figure 6.
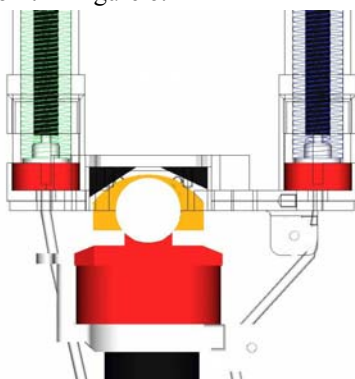

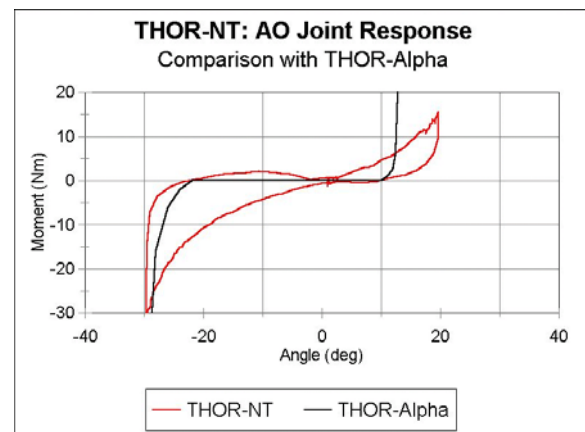**Figure 5. Cam-stop mechanism in THOR NT OC joint.**

Figure 6. Comparison of response at OC in THOR Alpha and NT.

The new joint makes the response more biofidelic, with gradually increasing resistance in both flexion and extension. The resistance is controlled by the engagement of a brass cam against shaped rubber stops fore and aft of the cam. The range-of-motion at the A-O joint has been increased to 20° in flexion and 30° in extension.

A problem that was sometimes encountered with the Alpha neck was variability in the output of the uniaxial load cells when loading was not completely normal to the load cell surface. A special, miniature, uniaxial load cell from R.A. Denton (Model: 6005) is now used in this location. It has been tested and found to be less sensitive to off-axis loading.

Another response issue with the neck that was addressed in the NT modifications was the possibility of hard bottoming of the die springs that are used within the head to provide for the simulation of neck musculature. The spring-only design in the Alpha was modified to include a Tygon tube insert within the front spring and a Viton insert within the rear spring. This provided a more gradual increase in the forces generated by the spring when it was being compressed near to its limit. The strength of the die spring was modified such that the combined spring/tube stiffness would be comparable to the original spring. Figure 7 shows a comparison of the behavior of the old and new designs when bottoming does occur.
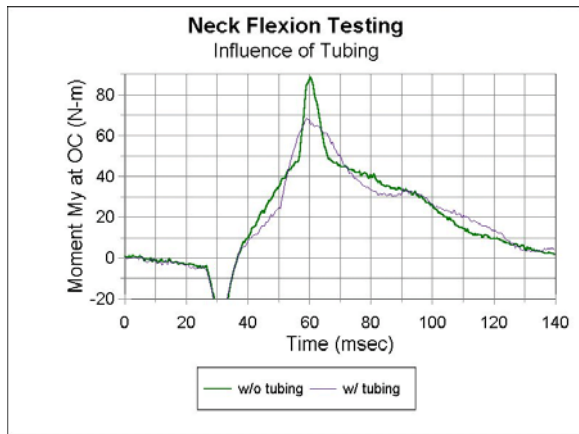
Figure 7. Reduction of bottoming effect of single spring by adding rubber tube insert.

Shoulder

There were a number of modifications made in the shoulder assembly of the THOR Alpha. These include:

- Changing the shape and material of the shoulder pads with improved anthropometry
- More robust shoulder stops
- Modified clavicle

The new shoulder pads are made from Monothane, a one-part polyurethane resin with improved durability characteristics over the two-part Urethane used in the Alpha shoulder pads. During this process, the anthropometry of the shoulder was also improved, with the shape of the new pad being obtained from the 3D representation of the 50th percentile male AATD form. Figure 8 shows a picture of the new pad on the dummy's right shoulder and the old Alpha pad on the left.
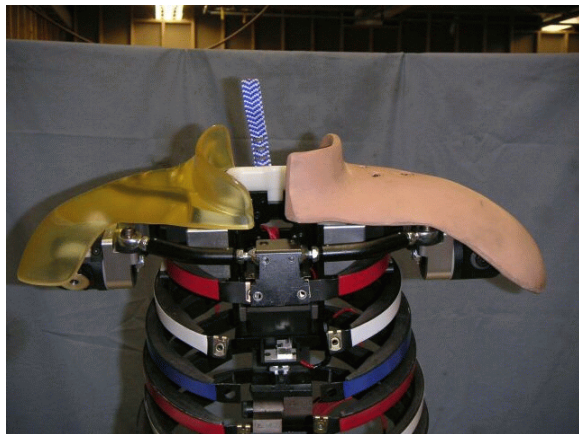

Figure 8. Comparison of new NT shoulder pad on left and original Alpha pad on right.

The stops that controlled the shoulder fore and aft motion were also modified and made more robust. The stops in the Alpha were attached using bolts that ran through the rubber material, which sometime resulted in the bending of the bolts in overload situations. The shoulder stops in the new NT design are glued to metal plates which are directly fastened to the spine. This design also improves the durability of the stops themselves with the elimination of the rather large bolt holes which had weakened the original structure. The new design increases the clearance between rib #1 and the shoulder block, where contact may occur under severe loading conditions. In addition, to simplify the assembly of the shoulder, the rib shelf, which was a separate structure, is now welded to the shoulder block. Figures 9 and 10 show the difference in the designs of the Alpha and the NT.
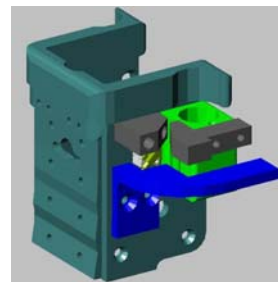

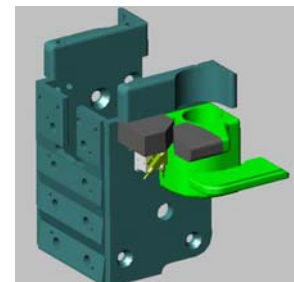
Figure 9. Shoulder block design in THOR Alpha

Figure 10. Shoulder block design in THOR NT.

Small modifications were also made to the clavicle. This included reducing the bend angle of the clavicle from 27° to 10°. This was meant to improve the relative distribution of belt load between the shoulder and the ribcage. The second modification was in adding a small, spherical cap to the distal end of the clavicle to improve belt retention. The cap was made to be approximately compatible with the general clavicle geometry seen in the human.

Spine

The improvements in the spine included:

- Changing to an injection molded process for making the thoracic and lumbar flex joints
- Removing protrusions at the rear of the ribs which were being used for routing of the cables
- Easier assembly of the T1 triaxial accelerometer
- Clearer markings for the spine pitch change mechanism

The bonding process used for making the two flex joints in the Alpha was changed to an injection molding process to improve the durability of the components. The new parts, like the NT neck, are also made of Butyl rubber, instead of Urethane, but the bending stiffness of the earlier design was maintained. The flex joints were tested in both static and dynamic conditions to ensure that the units were durable and repeatable over a minimum of thirty tests.

A number of smaller modifications were made to make the assembly of the spine easier and the appearance cleaner. The rear of the spine in the NT was improved by removing the wire routing clamps that were used in the Alpha. There is easier access to the T1 triaxial accelerometer block, which makes assembly and disassembly easier. Clear markings were added to the pitch change mechanism that sits at the bottom of the thoracic spine and allows the dummy to be put in four different sitting configurations.

Thorax

The changes to the thorax included:

- Changing to a Monothane bib
- Adding locating pins on the spine for attaching the ribs
- Reducing rib steel impingement on the damping material
- Modified sternum to improve frontal sternal impact response
- Modified rib #1 to improve frontal sternal impact
- Improvement in the shape of the jacket for better interface with the seat back

The new bib in the NT is shown in Figure 11.

Figure 11. Design of THOR NT Monothane bib.

The Urethane bib used in the Alpha was susceptible to damage under high belt loads. The Monothane was found to be more durable under these conditions.

The bib was especially shaped to be thicker in its lower part, where it covered the bottom three ribs. Testing at UVa had indicated that the lower ribcage in THOR Alpha was more decoupled from the rest of the ribcage, than expected for a human subject. In order to improve the coupling, the lower part was made thicker.

The addition of the locating pins to properly retain the ribs to the spine was based on comments from the FID consortium who had noticed that the ribs could be improperly assembled if care was not taken. The addition of the pins eliminates any variability during assembly.

GESAC had also observed during post-test inspections of the ribs that for certain test conditions the steel edge of a rib could load the damping material of the rib immediately below it leading to gouging. To reduce this effect, the damping material height of the lower three ribs was increased by about 1.5 mm so that contact would be between the damping material layers rather than steel with the damping material.

Both the mid-sternal mass and rib #1 were modified in order to improve the response of the thorax to sternal impact. Figures 12 and 13 show the difference between the sternal masses in the Alpha and the NT.

Figure 12. THOR Alpha mid-sternum.   **Figure 13. THOR NT mid-sternum.**

The Kroell tests at 4.3 m/s and 6.7 m/s are used to establish the biofidelity of the thorax under impact loading. During the original THOR development, the objective was to produce optimum response at 6.7 m/s while still ensuring that the dummy would be biofidelic at 4.3 m/s. This resulted in the dummy producing deflections at the lower end of the corridor for lower speed impact. NHTSA directed GESAC to make modifications which would result in greater deflection at the lower speed. This was motivated by

analyses of test results which indicated that typical loading rates seen in most accidents with newer cars were closer to the lower Kroell impact speed.

The jacket was also modified to provide a more rounded contour at the back and a better fit over the ribcage in the front.

Pelvis

A major revision of the THOR Alpha pelvis was done to improve both the anthropometry and the durability of the segment. The changes included:

• The external shape conformed closer to the AATD shape but modified to represent an undeformed configuration in order to allow the interaction of the seat and the pelvis flesh to generate the deformation
• The skin was made of PVC rather than Urethane
• The skeletal portion of the pelvis was made of modular components instead of single cast piece

The THOR Alpha pelvis had been made of Urethane and based on a standard sitting configuration. The NT pelvis skin is made of PVC which is more durable than the Urethane and a thinner skin can be used. The external shape is based on the 50th percentile male AATD form, but instead of the deformed shape of the pelvis skin, a somewhat undeformed shape was used. Figure 14 shows a 3D CAD drawing with the undeformed section shown on the left side (in red) and the original deformed section on the right side (in yellow).
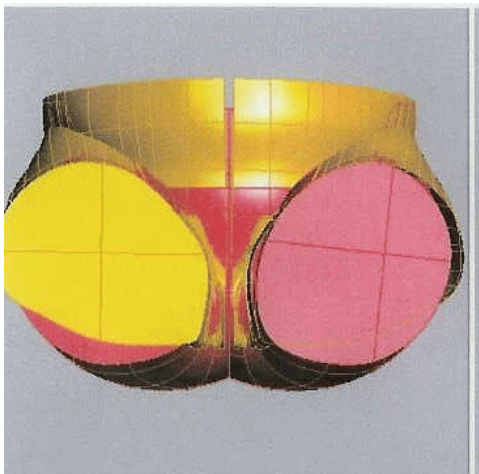

Figure 14. Comparison of undeformed and deformed pelvis sections.

Moving to an undeformed shape for the pelvis was motivated by the desire to generate the actual deformation through the interaction of the seat and the pelvis flesh. Thus the geometry of the external pelvis surface and the external femur surface (described below) was modified from the way it appears in the AATD form. Approximate undeformed measurements were taken from data obtained by McConville [1980], and from stereophotometric data obtained from 3D scans that were obtained at Wright-Patterson.

Average stiffness of the pelvis flesh was made to correlate with stiffness that has been estimated by Dabnichki, et al [1994]. The approximate compression of an adult sitting on a rigid seat was estimated to be about 20 mm. The objective in setting the flesh stiffness was to get the correct eye position when the dummy is positioned on the AATD hard seat.

The skeletal portion of the pelvis was constructed of a number of modular elements. The iliac wings were obtained by initially rapid prototyping the corresponding 3D CAD forms obtained from digitizing the original pelvis shape developed by Reynolds for a 50th percentile male [Reynolds, 1982] which served as the basis for the prototype THOR pelvis. The remaining components were machined parts which provides for symmetry, though this process adds a few more steps in the assembly of the dummy. It was also necessary to increase the mass of the pelvis over the THOR Alpha part, which required some of the components to be made of steel rather than aluminum.

Figures 15 and 16 show the skin and skeletal components of the new pelvis.

Figure 15. Vinyl pelvis skin of THOR NT.

Figure 16. Front view of THOR NT pelvis (skeletal).

Femur

The changes made to the THOR Alpha femur are:

*   Increased length of femur to match AATD data (hip joint - knee joint length)
*   Changed femur skin to PVC with a zipper for easier access to the femur
*   Added a representation of the trochanter
*   Changed the femur neck from a cast part to a machined part

The THOR Alpha femur, as measured between the hip joint and knee joint, was approximately 20mm shorter than the AATD. This length has been corrected in the NT. A Delrin part has been fastened to the femur neck region to represent the trochanter. Figure 17 shows the skeletal portion of the modified femur.

Figure 17. THOR NT femur (with trochanter).

The femur skin was also modeled from the seated 50th percentile male AATD form, but with the contour in an approximate undeformed shape, in a manner similar to what was done for the pelvis skin described in the previous section. The contour of the femur was made compatible with the distal section of the pelvis contour. A zipper was added to the skin to allow for easier access to the internal femur structure (which was being included in the Alpha skins as well).

Lower Leg and Foot

Some small modifications were made to the THOR-Lx. These include:

*   Pin added to ensure proper placement of the tibia tube relative to the upper tibia load cell
*   Clearance added for the tibia puck fasteners to ensure no binding occurs when the puck is compressed
*   Improved retention of the foot skin to the foot plate

Other Improvements

There were some general improvements made to the dummy and its handling as given below:

*   Common electric ground was achieved by maintaining electrical continuity between segments which may lose continuity during a test such as across the neck. This should prove useful in tests where large static electric charge may be generated, such as airbag tests.
*   New uniaxial tilt sensors replaced the biaxial ones used in THOR Alpha. These provide a linear response to angular change (the original ones had a non-linear dependence to angle)'
*   In the area of handling, a more accessible lifting point is now used for lifting and moving the dummy. A new H-point tool has also been designed which only needs access to the pelvis, rather than the old design which had to be attached to both the femur and knee joint.

THOR NT CERTIFICATION

Some of the modifications, such as in the head, neck, and thorax required selected certification tests to be undertaken to ensure that the THOR NT would meet the original biofidelity requirements of the THOR dummy. The following describes the results for the

selected certifications tests.

Head

A new certification test was added to evaluate the impact response of the head. Some of the laboratories indicated that they would prefer an isolated head drop test, similar to that done with the Hybrid III head. The new, integrated head skin was designed to allow for both the head impact certification test similar to that described in the THOR Certification Manual [GESAC, 2001b], and the head drop test. The requirement for the head drop test was the same as that for the 50th percentile Hybrid III, i.e. for a forehead impact from a height of 376 mm, the resultant head acceleration should be within the range of 225 g - 275 g. Figures 18 and 19 show the response for both the original whole body head impact test and the head drop test.
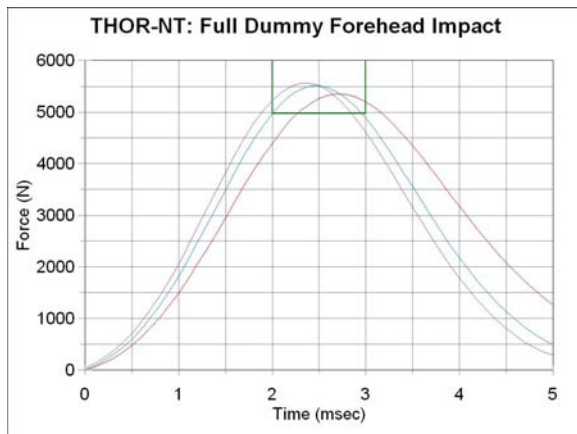
Figure 18. Response of THOR NT (whole body) to frontal head impact.
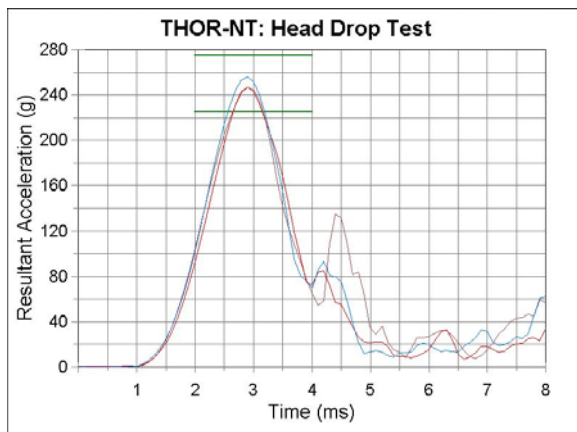

Figure 19. Response of THOR NT head to isolated head drop.

In the head drop test, a secondary impact was seen that was sometimes larger than seen with the Hybrid III head. This was attributed to the facial structure just below the forehead, which is significantly different from the aluminum face in the Hybrid III. The response from the whole body head impact was also considered to be acceptable and similar to that of THOR Alpha.

Face

During initial testing of the face with the new, integrated skin, it was found that the rod impact generated lower than expected peak force and some tuning of the skin thickness had to be done around the area of the eyes to increase the force. After this modification, both the disk and rod impacts were successfully performed on the new head. The results are shown in Figures 20 and 21.
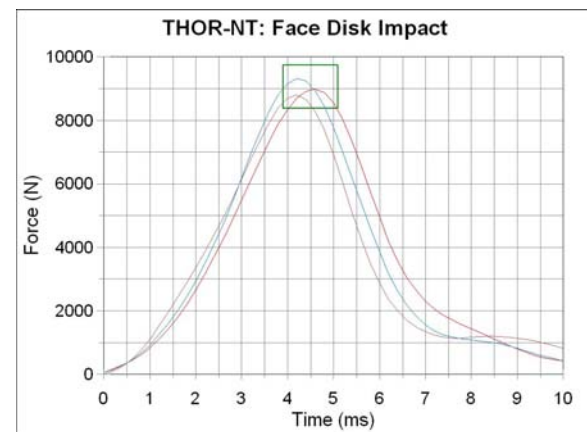

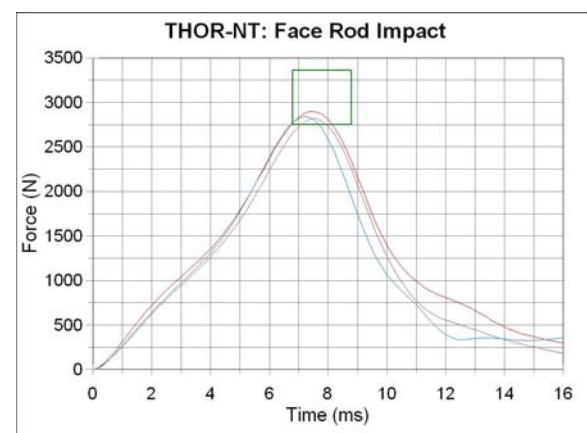**Figure 20. THOR NT face response to disk impact.**

Figure 21. THOR NT face response to rod impact.

Response of the new face to both impact conditions were seen to be fairly similar to that of the THOR Alpha.

Neck

As pointed out in the discussion of the new, injection molding process used to build the THOR NT neck, a different material, Butyl rubber, was being used. A fairly lengthy series of tests were conducted to ensure that the response of the neck had not changed significantly, and that the durability, repeatability, and reproducibility were improved. As mentioned earlier, a large number of repeat tests were conducted on a single neck. Figures 22-25 show the responses of one neck at two points in its test cycle in both the frontal flexion and lateral flexion tests.
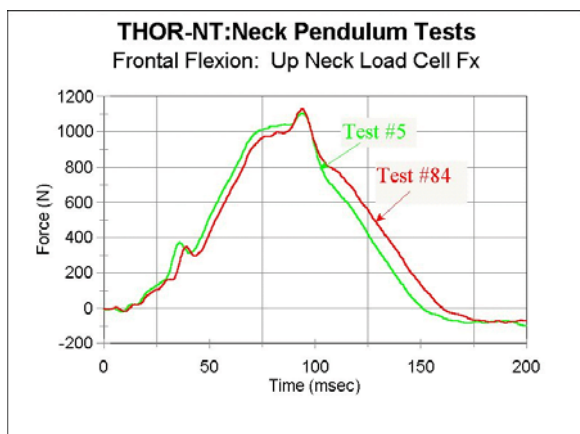

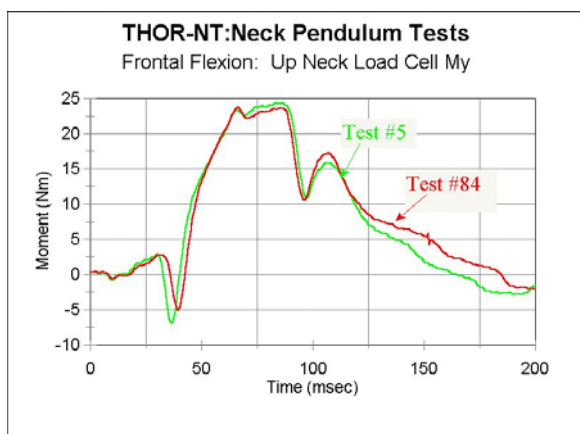**Figure 22. Repeatability of THOR NT neck Fx in frontal flexion.**

Figure 23. Repeatability of THOR NT neck My in frontal flexion
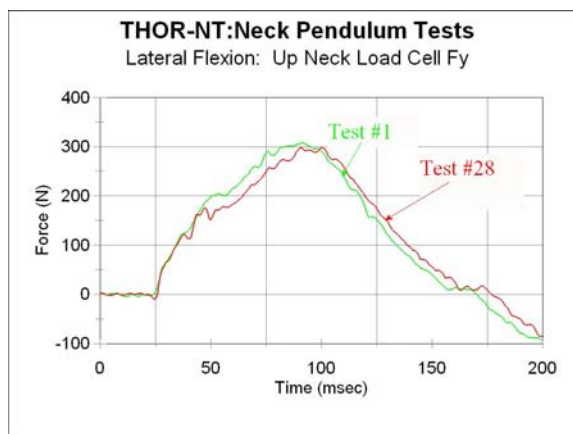

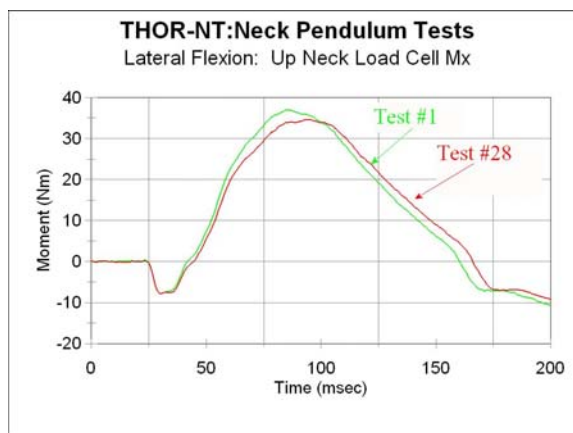**Figure 24. Repeatability of THOR NT neck Fy in lateral flexion.**

Figure 25. Repeatability of THOR NT neck Mx in lateral flexion.

It is seen that in both frontal and lateral response, good repeatability is maintained even after multiple tests.

Another important property of the new, molded necks that was evaluated was their reproducibility. Figures 26-28 show the variation in response of three different necks in frontal flexion, extension, and lateral flexion.
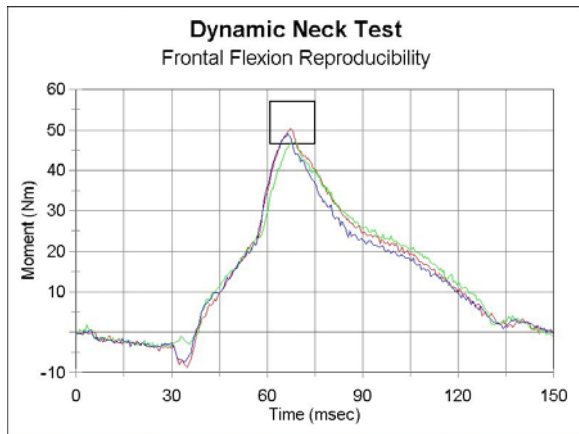
Figure 26. Reproducibility of three THOR NT necks in frontal flexion (total OC moment).
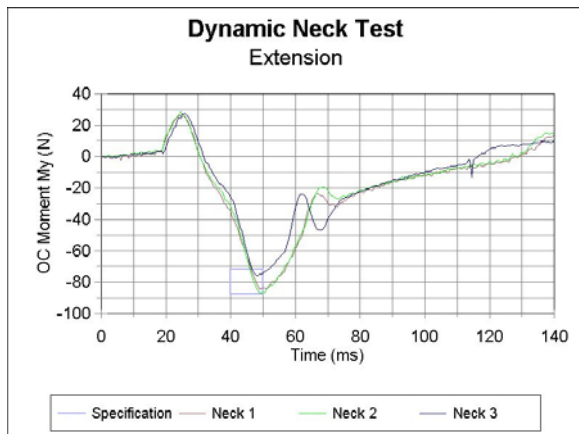

Figure 27. Reproducibility of three THOR NT necks in extension (total OC moment).
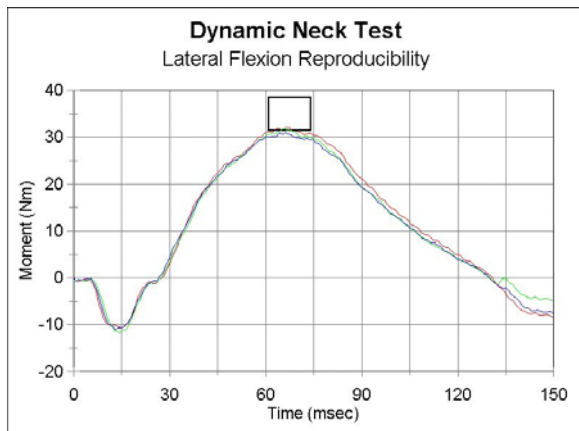

Figure 28. Reproducibility of three THOR NT necks in lateral flexion (total OC moment).

Responses of the three necks in all three directions are seen to be reproducible, though in extension some variability is seen. The responses are also close to the corridors that were developed earlier for the THOR Alpha neck (denoted by the rectangles in the above graphs). Though the NBDL type of sled tests have not been conducted with the new necks, the fact that their responses are similar to the original bonded necks, which did meet or were close to the NBDL kinematic corridors, makes it likely that the new necks will also meet the original biofidelity requirements.

Thorax

The final area where the modifications had been made that could influence its biomechanical response was in the thorax. As mentioned in the discussion of the design modifications made in the NT, they were partly motivated by the need to improve the low speed Kroell response compared to that of THOR Alpha. Figure 29 shows the response at the lower 4.3 m/s impact speed.
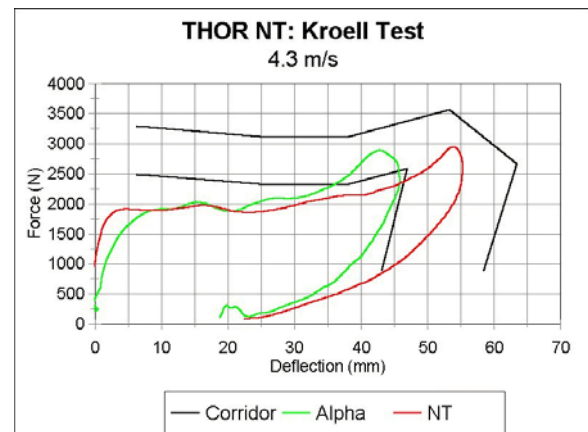

Figure 29. Comparison of THOR Alpha and NT Kroell impact response at 4.3 m/s.

It is seen that there has been a significant increase in the deflection, though no significant change in the peak force. Figure 30 shows the response at the higher 6.7 m/s impact speed.
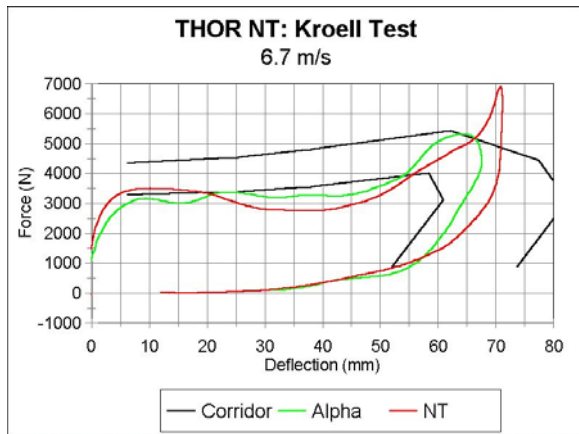
Figure 30. Comparison of THOR Alpha and NT Kroell impact response at 6.7 m/s.

In this case, it seen that peak force at the end shows a bottoming effect and results in the force exceeding the Kroell corridor. Since, the aim of the modification was to improve the response at the lower speed, and the higher impact speed was now considered something of an overload situation, the higher peak force was considered acceptable.

During the certification tests for the THOR NT, small adjustments were made to the corridors developed for the THOR Alpha for the head and face impacts, the neck response, and the Kroell tests. All the modifications were within or close to the biofidelity corridors as given in the THOR Biofidelity Manual [GESAC, 2001a].

CONCLUSION

A significant number of changes have been made to THOR Alpha in the areas of durability, biofidelity, anthropometry, and ease of use, to arrive at the NT design. The revised NHTSA Frontal Impact Dummy - THOR NT - is shown in oblique view in Figure 31.

Figure 31. View of fully assembled THOR NT dummy.

Areas of improvement include:

*   Improved Durability: The components that were particularly susceptible to failure in the THOR Alpha have been modified. Among these, major improvements were made to the neck and the two flexible joints by going to an injection molded process for manufacturing the parts. The pelvic and femurs skins are now made of PVC which provide significantly greater tear strength than the older Urethane design. The material for the shoulder pads and bib were changed to Monothane which has superior tear and abrasion resistance compared to the older Urethane materials. In the head, a single, integrated head skin is used instead of separate skull and face skins in the Alpha. This provides greater protection to the face Confor foam inside. Finally, a number of smaller changes, e.g. in the shoulder stops, have been made to improve the durability of these parts.
*   Improved Anthropometry: The head/neck interface geometry has been improved to provide a smoother surface for engagement with airbags. The shoulder pads have been designed to follow the external shape of the 50th percentile male AATD shoulder. The external shape of the pelvis and femur skins have also been designed based on the AATD form, but modified to represent the flesh in an undeformed state, so that it can better

represent the interaction with a vehicle seat. The femur length has been corrected and a proper representation of the trochanter has also been added.

• Biofidelity: The mechanism at the A-O joint of the neck has been changed to allow for a more biofidelic response with continuously increasing resistance in flexion and extension and the range-of-motion in both directions has been increased. Parts of the ribcage and sternum have been redesigned to improve the sternal impact response in the low speed Kroell test.

• Ease of Use: Locating pins in the spine allow for easy assembly of the ribs. A similar pin in the tibia allows for proper alignment of the tibia tube. Grounding straps provide electrical continuity across the dummy.

Certification tests for the head, face, neck, and thorax were performed to ensure that the modifications did not lead to any significant change in response of the NT as compared to the Alpha. Updated certification corridors have been developed for selected tests. A head drop certification test has also been added to verify proper response characteristics of the forehead in impact. An extended series of tests were performed to evaluate the durability and response of the new molded neck, and the testing confirmed that the new necks were repeatable and reproducible.

The THOR NT has addressed many of the limitations that were noted with the THOR Alpha dummy. The next phase of the evaluation will involve testing in different sled test configurations; a number of laboratories already have or are in the process of testing the NT. Results from these tests will be used to evaluate the performance of the dummy in real-world situations.

ACKNOWLEDGMENTS

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DESIGN OF A BIOFIDELIC INSTRUMENTED 3.4 KG INFANT DUMMY

Y. Wang
N. Rangarajan
T. Shams
GESAC, Inc.
United States
T. Fukuda, T. Yamada
Aprica Kassai, Inc.
Japan
C. Jenny
Brown University
United States
Paper Number 05-0456

ABSTRACT

A new infant dummy has been designed, manufactured, and tested, representing an average newborn of mass 3.4 kg. This dummy is a successor of the 2.5 kg newborn dummy developed earlier which had represent at 10th percentile Japanese newborn. Gross data such as total weight, length, and head circumference were taken from several sources including the Centers for Disease Control [CDC, 2000]. More detailed measurements were obtained from newborns in two Japanese clinics. Dynamic response data for head, neck, thorax, and abdomen were defined by scaling adult data. The dummy has 11 segments (head, neck, torso, upper arm, lower arm and hand, upper leg, lower leg and foot). The torso is further divided into shoulder, chest, abdomen, and pelvis, all connected to a flexible spine. Segments are connected by joints which provide human like range of motion. The dummy is instrumented with 26 sensors, including triaxial accelerometers at the head CG, upper and lower neck, thorax CG and pelvis CG; 3-axis angular velocity sensor in the head; uniaxial load cells in the neck and lumbar spine; string potentiometer to measure chest deflection; and five force sensors on the abdomen. This paper describes the methodology used to develop the design and the results from biofidelity testing.

INTRODUCTION

In late 2000, Aprica Childcare Institute funded GESAC, Inc to design and develop a new, small, infant dummy, which is now known as the Aprica 2.5 kg infant dummy. The development of the infant dummy was motivated by the need to have a more biofidelic and instrumented dummy which would represent a small infant for evaluating restraint systems. Aprica had used the TNO P0 dummy for such evaluation, but it was felt that the dummy was unsuitable because:

1. Its weight was over the Aprica target of 2.5 kg, the weight of a 5% percentile Japanese newborn.
2. The neck was too stiff. It is known that the neck of the newborn is unstable and generally unable to support the head. The P0 dummy appears to model the instability of the neck by integrating a nearly unconstrained A-O joint in the dummy. The head pitches around the A-O joint easily but without any involvement of the neck.
3. The dummy was not instrumented.

An overview of the 2.5 kg infant dummy is given by Rangarajan [2002]. Following the successful testing of the 2.5 kg infant dummy, Aprica funded GESAC to modify the design of 2.5 kg infant dummy to represent a 50th percentile newborn with a mass of about 3.4 kg. The objective of the new dummy, apart from representing a wider range of infants, was to improve its biofidelity and also to enhance its instrumentation.

ANTHROPOMETRIC DATA

The anthropometric data of an average infant were collected from several sources including the CDC for the overall height, weight and head size, the Hirokawa and Nishikawa Clinics for measurements on actual subjects, and a 1975 report on anthropometry of US infants and children [UMTRI, 1975]. Scaling from data of the next closest age group was also employed

to determine values of certain dimensions when measured data were not available from existing sources. Some major target dimensions of the 3.4 kg infant dummy are listed in Table 1. Mass distribution of the dummy is listed in Table 2. The actual values are also listed in the two tables for comparison.

Table 1.
Key target and actual dimensions of the 3.4 kg infant dummy.

Parameters	Target (mm)	Actual (mm)
Overall height	504	530
Upper arm length (shoulder to	77	94
Lower arm length (elbow to wrist)	74	70
Hand length	62	38*
Upper leg length (hip to knee)	89	115
Lower leg length (knee to ankle)	86	95
Leg length (knee to heel)	120	123
Foot length	76	67
Foot breadth	32	38
Mid-thigh circumference	160	170
Leg at knee circumference	135	130
Calf circumference	125	120
Ankle circumference	87	90
Head circumference	348	352
Head depth	117	115
Head breadth	94	98
Head height (head top to chin)	130	108
Neck length (OC to C7/T1)	60	66
Mid-neck circumference	200	195
Shoulder breadth	158	160
Shoulder to crotch length	218	225
Shoulder circumference	360	310
Chest breadth	109	106
Chest depth	84	74
Chest circumference	340	310
Waist breadth	105	106
Waist depth	90	91
Waist circumference	335	334
Hip breadth	113	115
Hip circumference	338	295
Spine length (T1 to L5)	179	180

* dummy's hands folded

Table 2.
Target and actual mass distribution of the 3.4 kg infant dummy.

Segment mass	Target (g)	Actual (g)
Head	1067	999
Neck	168	79
Torso (chest, abdomen, pelvis)	1229	1204
Upper arm	77	83
Lower arm and hand	59	92
Upper leg	206	192
Lower leg and foot	126	152
Total mass	3400	3320

INFANT DUMMY DESIGN

Figure 1 shows a picture of the 3.4 kg infant dummy in its sitting posture with instrumentation wires at the side. The dummy consists of 11 segments, i.e., the head, neck, torso (which includes chest, abdomen and pelvis), upper arms, lower arms with hands, upper legs, and lower legs with feet.

Figure 1. The 3.4 kg infant dummy with instrumentation wires.

The head has an inner aluminum housing which provides room for instrumentation of the head. The metal housing is then covered by a specially formulated urethane skin. Figure 2 shows the metal housing and the cap of the head.

Figure 2. Metal housing and cap of the infant head.

The spine structure consists of the neck, thoracic spine, and lumbar spine. The shoulder and pelvis assemblies are integrated within the spine structure, along with instrumentation, as shown in Figure 3. The neck is connected to the head at the Occipital Condyle (OC) with a pin joint. The neck is molded as a urethane column with reinforcement at the center. Its lower end is attached to the shoulder block through a uniaxial load cell.

Figure 3. Neck-spine assembly with shoulder, chest block and pelvis. Wires are of two uniaxial load cells and a string potentiometer.

The shoulder block is machined of Delrin plastic. Two spherical ends made of aluminum serve as the shoulder joints and are attached to it. The shoulder block is fixed to the chest block which houses a string potentiometer.

Between the chest block and the pelvis there is a flexible lumbar joint. The lumbar joint is molded as a urethane column with reinforcement at the center. A uniaxial load cell is also attached to the lower end of the lumbar joint.

The pelvis is also made of Delrin and sits under the lumbar load cell. The hip joints are attached under the pelvis block. Like the shoulder joints, these consists of spherical ends made of aluminum.

The upper leg is connected to the pelvis by the ball and socket hip joint. The friction of the joint can be adjusted by tightening or loosening a screw attached to the joint for this purpose. The friction can be set to the standard 1G level by this method. The lower leg is connected to the upper leg by a pin joint. Its friction can also be adjusted. The femur and tibia bones are made of ABS plastic. A specially formulated layer of Urethane is molded outside the bones to simulate the flesh and skin.

The upper arm is connected to the shoulder by a ball and socket joint. The friction of the joint can be adjusted by tightening or loosening a screw attached to the joint for this purpose. The lower arm is connected to the upper arm by a pin joint. Its friction can also be adjusted. The humerus and forearm bones are made of ABS plastic. A specially formulated layer of Urethane is molded outside the bones to simulate the flesh and skin.

Figure 4. A prototype of the rib cage.

The rib cage is made of a polycarbonate outer shell lined with strips of damping material. An aluminum mass is attached to the front of the cage to simulate the sternum. Figure 4 shows a prototype of the rib cage (the final design is dimensionally different from the one shown here, but structurally the same). The actual rib cage can be seen in Figure 5.

The torso flesh/skin is molded as one piece of specially formulated Urethane. A slit runs along the center of the back of the flesh/skin to provide access to internal parts. A zipper is sown to the flesh/skin at the slit. Figure 5 shows the torso and its inside structure.
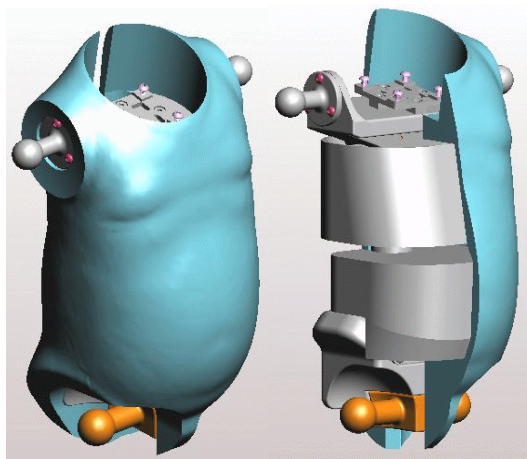
Figure 5. The torso of the dummy.

The abdomen of the dummy is filled with a piece of foam with a stiffness that is tuned to meet the abdomen biofidelity requirements.

INSTRUMENTATION OF THE INFANT DUMMY

The goal of the instrumentation with this dummy is to provide as many channels as possible for various testing conditions. This dummy is expected to be used in the evaluations of child restraint system, comfort of strollers, as well as in the study of shaken baby syndrome. A deliberate design effort was made to meet this goal. The 3.4 kg infant dummy is able to measure a maximum 26 channels of data. A full list of the instrumentation is shown in Table 3.

**Table 3.
Instrumentation of the 3.4 kg infant dummy.**

Location	Sensor type	Channels
Head CG	Angular velocity	3
Head CG	Accelerometer	3
Upper neck	Accelerometer	3
Lower neck	Accelerometer	3
Lower neck	Load cell	1
Thorax CG	Accelerometer	3
Thorax	String potentiometer	1
Lower lumbar end	Load cell	1
Pelvis CG	Accelerometer	3
Abdomen	Flexible Force	5

Angular acceleration is assumed to be an important dynamic variable which correlates with possible injury to the head [Duhaime, et al, 1987], and may be more important than simple linear acceleration. Thus the capability of measuring angular accelerations was considered to be an important instrumentation requirement. The angular velocity sensor model ARS-06S by ATA Sensors was used. The workable frequency range of these sensors is 0.4 Hz ~ 1.0 kHz, which is deemed suitable for most impact and shaken testing. Figure 6 shows the triaxial angular sensor mounted inside the head housing.

Figure 6. Triaxial angular velocity sensor housed inside the head.

The accelerometer model ASM-200BA by Kyowa Electronic Instruments was used for all acceleration measurements. This model is compact, has a capacity of 200 g, and offers high resolution. It is suitable for all testing conditions currently planned for the infant dummy.

The uniaxial load cell model 6398 by Robert A. Denton, Inc. was used for the axial load measurements at the lower neck and the lower lumbar spine. This load cell has a capacity of 200 lbf (890N) and a compact size (35.0 x 35.0 x 7.6 mm). It was introduced to provide additional information on tension and compression forces at the top and bottom of the spine.

The displacement of the chest when chest compression occurs is also of interest in evaluating thoracic injury. A string potentiometer is installed at the position of about the 6th/7th thoracic spine to provide an estimate of chest compression. One end of the string is attached to the spine and the other end is connected to the sternum mass. The compact string potentiometer (model 174-0321TR) made by SpaceAge Control was used. It has a maximum travel of 1.5" (38 mm), and is small enough to be housed within the thoracic spine.

In testing the child seat, it is also desirable to know the forces applied to the abdomen area of the dummy due to belt or other interactions. Since the abdomen is very flexible and has very limited space for instrumentation, a structurally flexible measuring mechanism is desirable. To meet this requirement, FlexiForce sensor model A201 by Tekscan was used. These sensors are paper-thin, flexible, and have a capacity of 1~1000 lbf (4.4~4450 N). Five of these sensors of 100 lbf (445 N) capacity were affixed to a piece of cloth with an area corresponding to the face of the abdomen. This set of sensors is put between the molded flesh and the abdomen foam when it is necessary to measure the force sustained by the abdomen. The set of sensors can be taken out of the dummy when the measurement is not required. Figure 7 shows the set of FlexiForce sensors.
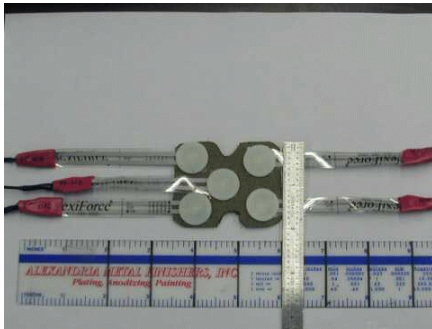

Figure 7. A set of 5 FlexiForce sensors used for abdomen force measurement.

BIOMECHANICAL REQUIREMENTS

Scaling

Since there are only very limited test data on newborn infants, scaling was employed to define the biomechanical requirements for the newborn. The biomechanical requirements for the 50th percentile male dummy were scaled using the following scaling factors to obtain the requirements for the newborn.

Three basic scaling factors:
$$\lambda_m = m_s / m_p \qquad (1).$$
$$\lambda_\rho = \rho_s / \rho_p \qquad (2).$$
$$\lambda_E = E_s / E_p \qquad (3).$$

where, λ is the scaling constant, m the mass, ρ the density, and E the modulus of elasticity, respectively. The subscript s refers to scaled data, and p to prototype or standard data. In this study, the mass density is assumed to be the same for adults and newborns, therefore, λ_ρ=1.0.

Other scaling factors can be obtained by using their mathematical relationships as follows.

Scaling factor for velocity:
$$\lambda_V = \sqrt{\lambda_E} \qquad (4).$$
Scaling factor for time:
$$\lambda_T = \lambda_L / \sqrt{\lambda_E} \qquad (5).$$
Scaling factor for acceleration:
$$\lambda_a = \lambda_E / \lambda_L = \lambda_E / \sqrt[3]{\lambda_m} \qquad (6).$$
Scaling factor for force:
$$\lambda_F = \lambda_L^2 \lambda_E = \lambda_m^{2/3} \lambda_E \qquad (7).$$

where λ_L is the length ratio between the scaled and prototype objects.

For the 3.4 kg infant dummy, the following biofidelity tests were defined: the head impact, the head drop test, thorax impact test, the abdomen impact test, and the neck pendulum test. The response requirements for these tests were scaled from the corresponding responses in the adult 50th perecentile dummy.

Head Impact Requirement

Head impact requirement was scaled from the 50th percentile male dummy requirement [Hodgson, et al., 1975]. Based on the scaling procedure, the impact at the forehead of the 3.4 kg dummy should be performed at a speed of 0.8 m/s using a rigid impactor of 6.04 kg. The impactor head has a shape of an oval of 50 mm x 75 mm. The scaled corridor for the force-time curve is shown in Figure 8.

In obtaining the scaling factor, λ_E, of the human skull, data from McPherson, et al., quoted by Melvin [1995] and Thibault et al. [1999] were averaged. Both studies had limitations in regard of sample sizes.
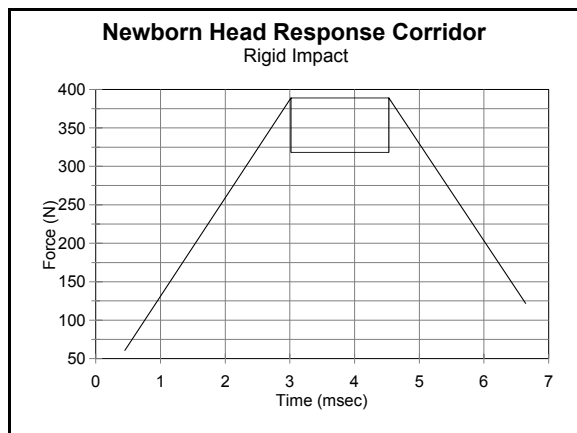
Figure 8. Biomechanical corridor for infant head impact.

Head Drop Test Requirement

Head drop requirement for the newborn was scaled from the 50th percentile male corridor [Hubbard, et al, 1974]. Using the scaling factors mentioned above, the head drop test of the 3.4 kg newborn should be performed at a height of 376 mm and should produce a peak resultant acceleration of 132~162 g.

A new study based on a small sample size by Prange et al. [2004] at Duke University has indicated lower peak acceleration for this test. These data can be incorporated into developing a new corridor in the future when more data are available. For the design of the current 3.4 infant dummy, the scaled corridor was used as a design target.

Thorax Impact Requirement

The thorax requirement for the 3.4 kg infant dummy was scaled from that for the 50th percentile male dummy [Neathery, 1974; Ratingen, et al, 1997]. Based on the aforementioned scaling method, the impact test for the 3.4 kg dummy thorax should be performed at a velocity of 3.3 m/s using a rigid impactor of diameter 50 mm with a mass of 1.1 kg. The force-deflection curve of the dummy at these conditions should be within the corridor shown in Figure 9.
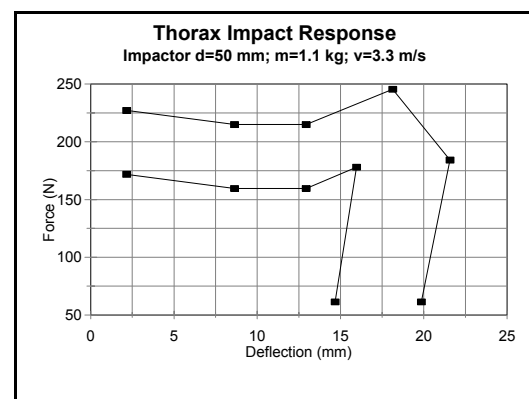

Figure 9. Biomechanical corridor for infant thorax impact.

Abdomen Impact Requirement

The biomechanical corridor for the abdomen impact of the 3.4 kg dummy was scaled from that of the 50th percentile male dummy [Cavanaugh, et al, 1986; Hardy, et al, 2001]. Based on the scaling method, the certification for the 3.4 kg dummy should be performed at a velocity of 4.7 m/s using a rigid rod impactor of diameter 10 mm with a mass of 1.4 kg. The force-deflection curve under these conditions should be within the corridor shown in Figure 10.
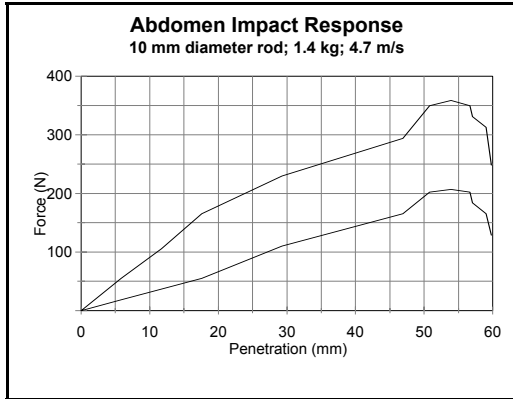
Abdomen Impact Response
10 mm diameter rod; 1.4 kg; 4.7 m/s

Figure 10. Biomechanical corridor for the infant abdomen impact.

Neck Pendulum Test Requirement

There are two kinds of requirements for the neck pendulum test, that is, the dynamic and kinematic requirements [Mertz, et al, 1971, Patrick, et al, 1976]. The dynamic requirement is characterized by a curve of the moment about the OC joint vs. the head rotation angle, and the kinematic requirement is a time history of head angle. Since the 3.4 kg infant dummy is not sufficiently instrumented to measure the OC moment, the current dummy design effort only focused on meeting the kinematic requirement.

The scaled kinematic requirement for the neck under the test pulses (see Figures 21 & 22) are shown in Table 4.

Table 4.
Kinematic requirement for neck pendulum test under the specified pulses.

Test	Peak angle (deg)	Peak time (ms)
Frontal	65 ~ 79	68 ~ 83
Lateral	36 ~ 46	63 ~ 77
Extension	-73 ~ -89	77 ~ 94

DISCUSSIONS

Anthropometric Resemblance

Comparing the target anthropometric data with the actual data of the dummy listed in Tables 1 and 2, it is clear that the design basically achieves its

goal to produce an anthropometrically human-like dummy. It provides a successful prototype for future improvement. Some dimensions and mass distributions can be further fine-tuned in later modifications. Since some target values are either estimates or scaled values from other age groups, the small difference between the target and actual values is deemed insignificant.

Biofidelity Certification Tests

The biofidelity tests described above, were performed to examine the performance of the designed infant dummy. Before the tests, the dummy was soaked in a temperature controlled room for 24 hours. The temperature was controlled between 69°F and 72°F.

The tests were carried out using a linear impactor shown in Figure 11 driven by a pendulum from one end. This impactor was designed to have the flexibility of attaching different moving masses and different impactor heads that are required by each test. Different speeds can be achieved by changing the drop height and the mass of the pendulum. The linear impactor was instrumented with an LVDT, a uniaxial load cell, and an accelerometer. The speed of the impactor was monitored by a velocity gate which is not shown in the picture. The speed gate can be seen in Figures 12, 16 and 18. During the tests, the impactor was fixed to a frame which also supported the driving pendulum and a height-adjustable platform for sitting the dummy.

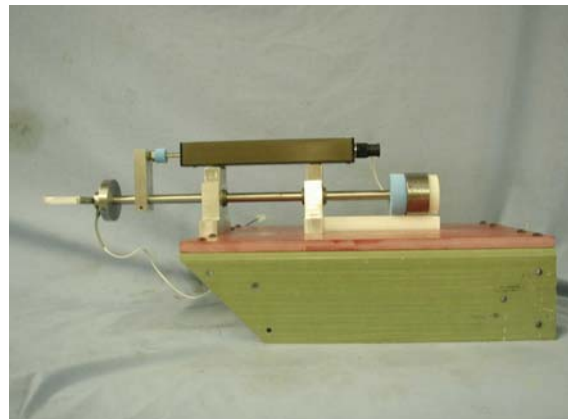The setups and results of each test are discussed below.

Figure 11. The linear impactor used for testing the infant dummy (with the impactor head for abdomen test).

Head Impact Test

Figure 12 shows the setup of the head impact test. The dummy was sitting upright on a hard plastic surface without any other support. The impactor was aimed at the center of the forehead of the dummy. The impact speed was 0.8 m/s.
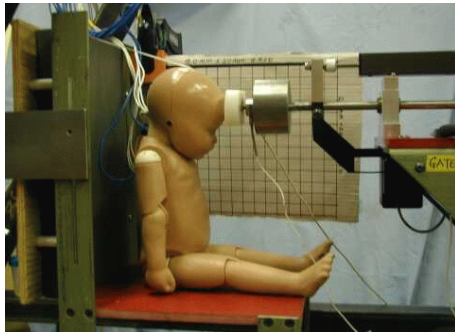

Figure 12. Setup of the head impact test.

Figure 13 shows the result of the head impact compared to its biomechanical corridor. It can be seen that the repeatability of the two tests were very good. The average peak force of the tests was within the scaled corridor, while the timing of the peak is about 1ms behind the scaled corridor. The performance of the head in the impact test was found to be acceptable.
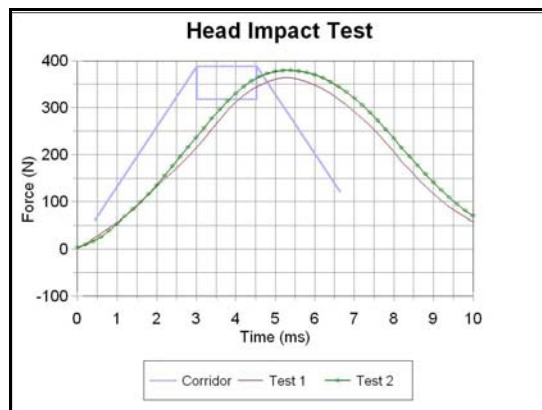
Figure 13. Head impact result of the infant dummy.

Head Drop Certification Test

The setup of the head drop test is shown in Figure 14. The setup is the same as used for testing the HIII dummy head. The head was dropped from a height of 376mm to a rigid surface (a thick steel

plate). Accelerations on three directions were measured to calculate the resultant value.

Figure 15 shows the drop test result. The tests showed good repeatability, and the average peak value of the three tests was within the scaled corridor.
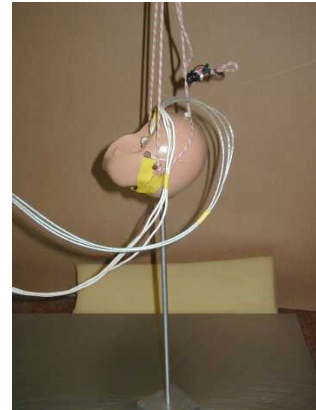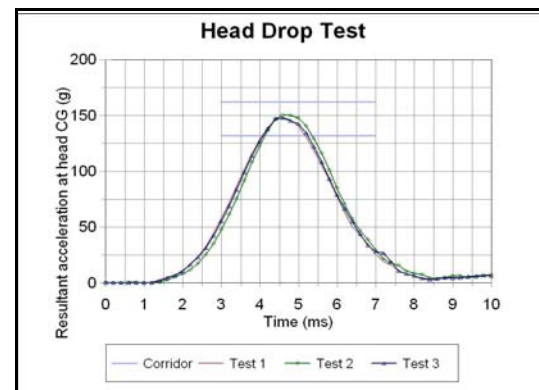

Figure 14. Setup for head drop test.

Figure 15. Head drop test result of the infant dummy.

Thorax Impact Certification Test

Figure 16 shows the setup of the thorax impact test. The dummy was seated so that the anterior chest is approximately parallel to the impactor head. The center of the impactor is aimed at the center of the chest, but avoiding impact with the shoulder block inside the dummy. The head of the dummy drooped naturally. The forearms of the dummy were raised to allow better view for the high speed camera.

Figure 17 shows the result of the thorax impact tests. It can be seen that the two tests repeated well. Though the force is close to the upper limit of the

scaled corridor, the maximum deflection of the chest is within the corridor. The design successfully achieved the relatively flat part of the force-deflection as in the scaled corridor, as well as, the hysteresis. The performance of the chest was considered to be acceptable.
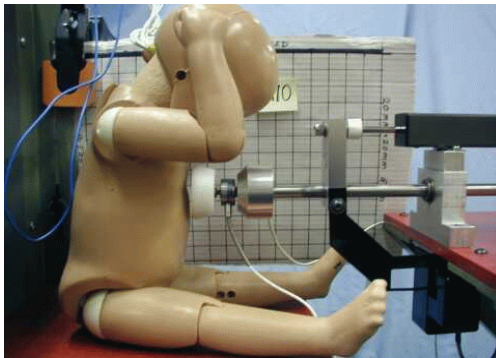

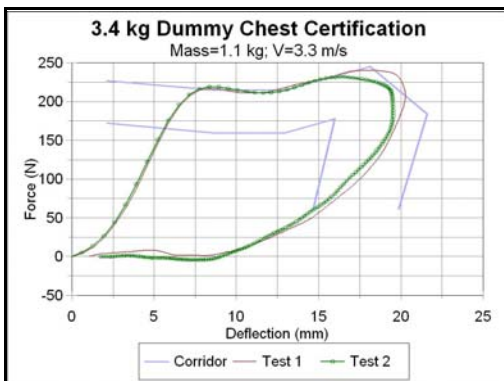**Figure 16. Setup of the thorax impact test (the forearms were raised to allow better high speed camera view).**

Figure 17. Test result of the thorax impact compared with scaled corridor.

Abdomen Impact Certification Test

Figure 18 shows the setup of the abdomen impact test. The dummy was seated with its torso upright. The impactor head was aimed at the center line of the abdomen. The arms were placed out of the way for better viewing by the high speed camera.

Figure 19 shows the result of the abdomen impact tests. The repeatability of the two tests was good. The abdomen was compressed by about 41mm, and the force during the compression fell

into the scaled corridor, which demonstrated that the current abdomen design was able to meet the target requirements.

Neck Pendulum Certification Test

The neck pendulum certification was performed in all three modes - frontal flexion, extension and lateral flexion. Currently the response of the neck is focused on the kinematic requirement. The angular motion of the head was calculated from the angular rate sensors located at the head CG. The time history of this curve was plotted against its requirement to see how the neck performs under different mode of dynamic loading. Neck axial load from the lower neck load cell and accelerations at head CG were also measured, but were not examined for response purposes.
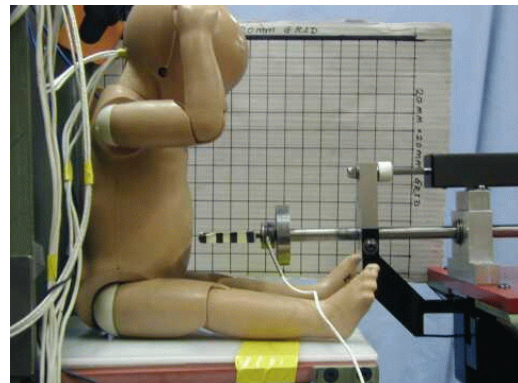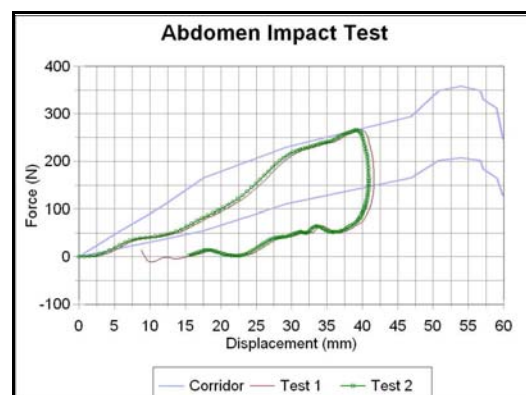

Figure 18. Setup of the abdomen impact test.

Figure 19. Test result of the abdomen impact compared with scaled corridor.

Figure 20 shows the setup of the neck pendulum test. The head-neck assembly was attached to the pendulum arm through an adapter. The drop angle of the pendulum was monitored by an angular

potentiometer at the rotation axle. The pendulum was stopped by a piece of foam attached to a rigid frame. The foam was carefully selected to produce the desired pulses for the testing.

Figures 21 is the pulse produced for frontal flexion and extension tests, and Figure 22 the pulse for the lateral test. For frontal flexion and extension, the pendulum was dropped from the same height, creating a pulse of about 25 g with a duration of about 20 ms. For lateral flexion, the pendulum was dropped from a lower height, creating a pulse of 12.6 g with a duration of 30 ms.

The results of the tests are shown in Figures 23, 24 and 25 for frontal flexion, extension and lateral flexion respectively. For the flexion test, it can be seen that the timing of the peak is close to the target kinematic requirement though the maximum head angle is slight higher. In the extension test, the maximum angle is also slightly higher than the required value, but the time of the peak is within the requirement. In the lateral test, the maximum angle is higher than the required value, but the time of peak is within the requirement.

Overall, the time of peak angles were within the kinematic requirement. The maximum angles were found to be slightly higher than the requirements. Since the requirements were based on scaling the adult 50th percentile neck responses on the pendulum, it is felt that additional biofidelity data should be developed before proceeding with any modifications to better tune the necks.
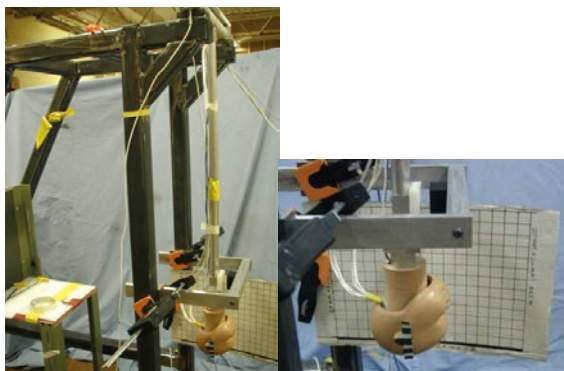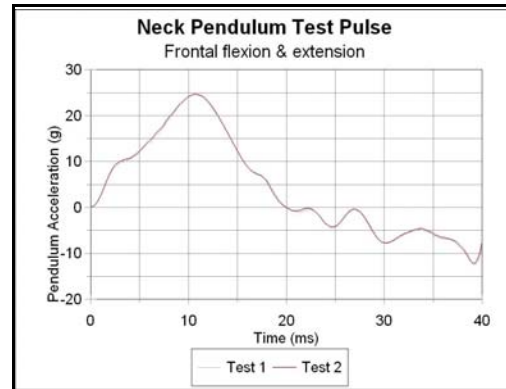

Figure 21. Acceleration pulse produced by the pendulum for frontal flexion and extension tests.
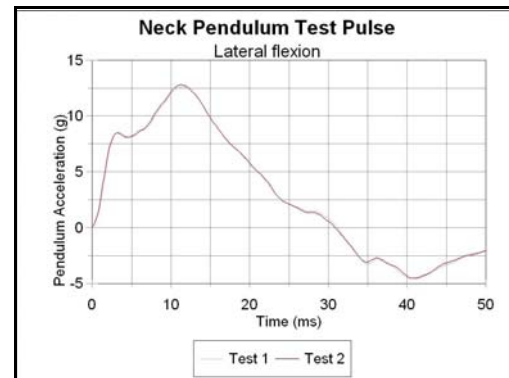
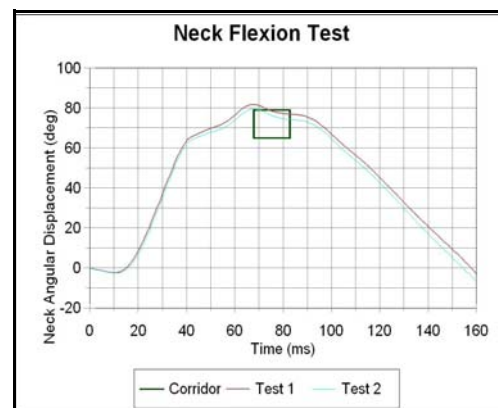**Figure 22. Acceleration pulse produced by the pendulum for lateral flexion test.**

Figure 20. Setup of the neck pendulum test for frontal flexion (left: overall view, right: local view).

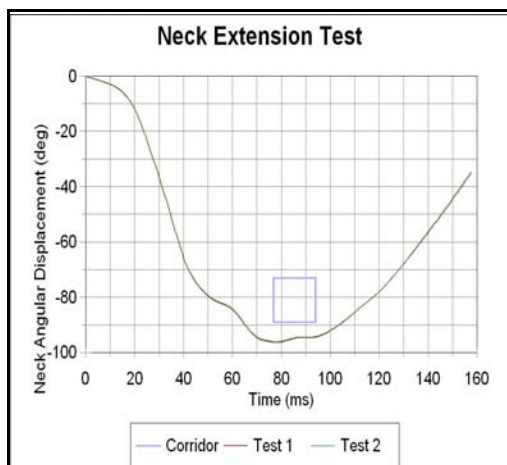**Figure 23. Time history of head angle during neck frontal flexion.**

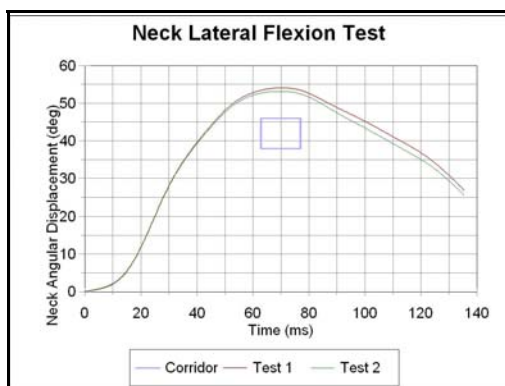**Figure 24. Time history of head angle during neck extension.**

Figure 25. Time history of head angle during neck lateral flexion.

CONCLUSION

A 3.4 kg infant dummy representing an average of newborn baby was developed. The dummy was designed to be used for various testing setups including child restraint system evaluation, stroller comfort evaluation, shaken baby syndrome studies, and others. The dummy includes 26 channels of data to provide sufficient data for the different tests planned using it. Biofidelity requirements for the dummy were obtained by scaling the requirements for the 50th percentile male dummy. The biofidelity tests have shown that the performance of the dummy either meet or are close to the target requirements.

As new data from cadaver testing on newborns come out in the future, some of the biofidelity requirement may need to be revised. However, the success with this dummy demonstrated that the design approach used this study can be employed to meet those possible revisions in the future.

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